Title: A novel approach to activate deep spinal muscles in space – results of a biomechanical model

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Running head: Lumbo-pelvic muscle training in space

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Abstract:

Introduction: Exposure to microgravity has various effects on the human musculoskeletal system. During spaceflight many astronauts experience low back pain and the risk of spine injuries is significantly greater post-flight. Nonetheless, the increased lumbo-pelvic injury risk is not specifically addressed by current countermeasures. Considering this, a novel exercise device has been developed to specifically counteract atrophy of deep spinal and postural muscles. The aim of the present study was to test the possibility of transferring this exercise concept from earth to space using a biomechanical simulation.

Methods: A biomechanical model of the exercise device was developed and validated using intramuscular electromyographic (EMG) data as previously acquired on a terrestrial prototype of the exercise device. The model was then modified to the needs of a 0-g environment, creating gravity-like conditions using shoulder straps.

Results: Modelled activation patterns of the investigated muscles were in line with the experimental data, showing a constant activation during exercise. The microgravity modifications of the model lead to increased muscle activation of deep spinal muscles and to decreased activation of superficial moment creating trunk muscles.

Discussion: The results of the biomechanical model suggest that the exercise concept can be transferred from 1-g to space conditions. The present study is a first step in the investigation process of a novel exercise concept and human studies should be conducted to confirm the present theoretical investigation.

Key words: low back pain, astronaut training, spinal health, lumbar multifidus, transversus abdominis, countermeasure

Introduction

The evolutionary process, as proposed by Charles Darwin [24] allowed the human body to successfully adapt to the 1-g environment of our planet Earth. When exposed to the microgravity environment in space, the human body experiences various immediate and long term adaptations that have, amongst other things, a significant impact on the structure and function of the musculoskeletal system [22].

Upright activities on earth cause the human body to be vertically loaded (Gz loading) and it has been found that the direction of loading is of particular importance for the integrity of the spine and its adjacent structures [4, 6, 25]. It has been well documented that, in the absence of Gz loading during space flight, or bed rest, the spine experiences structural and functional changes that make it more susceptible to injury when the Gz vector is restored [3, 5, 6, 23].

Interestingly, it has been observed that in a bed rest environment, some muscle groups of the lumbo-pelvic region adapt differently to others [4]. Under bed rest conditions the cross sectional area of deep spinal muscles and spinal extensor muscles decreases in size, however, the cross section of spinal flexor muscles increases [6]. Importantly, dysfunctional or atrophied muscles in the lumbo-pelvic region are directly linked to the prevalence of low back pain (LBP) [17]. Given the evidence suggesting that the functional and structural integrity of the transversus abdominis (TrA) and lumbar multifidus (LM) muscles are required to maintain spinal health [17, 18, 21], current rehabilitation regimens aim at specifically exercising these muscles [19, 27].

Current state of the art in-flight exercise interventions aim to counteract the physiological effects of microgravity by temporarily generating mechanical and metabolic stimuli to the human body. On average 1.75 hours of exercise (of a 2.5 hour daily allocation) is conducted each day, comprising typically of 1 hour of aerobic and 45 mins of resistive exercise [30]. Resistive exercise is done using the Advanced Resistive Exercise Device (ARED), a machine that allows performance of a various different exercises of the global movement muscles [30].

Aerobic on-board exercise is performed on either a treadmill or on a cycle ergometer [30]. Despite the fact that upright activities such as walking are associated with an activation of deep spinal muscles [7], evidence suggests that the currently applied treadmill exposure is not sufficient to preserve spinal health, given the four-fold increased risk of astronauts experiencing a herniated disc post flight [23].

Recently, a novel exercise device (the Functional Re-adaptive Exercise Device, or the FRED, Figure 1) has been developed [10] that showed promising results to immediately increase lumbo-pelvic stability during movement in healthy terrestrial individuals, as indicated by a decreased range of hip rotation in the transverse plane [14]. The device is similar in action to an elliptical trainer but requires the user to work against almost no resistance in order to move the feet. This feature constitutes a postural challenge to the user because of the need to actively control the movement of the legs while maintaining a stable trunk. It has been shown that the combination of weight bearing in an upright posture and functional leg movement while exercising on the FRED immediately increases the activation of TrA and LM compared to standing on an unstable base or leg movement without weight bearing [10]. Exercise on the FRED also appears to promote an immediate phasic-to-tonic shift in lumbo-pelvic muscle activation as well as an increase in spinal extensor and reduction in spinal flexor muscle activation [7]. These findings suggest that the FRED could provide a useful tool in the rehabilitation of astronauts [12].

Whilst previous findings support the use of the FRED on Earth, the aim of the present work was to evaluate the potential to use the FRED as a direct in-flight countermeasure to counteract atrophy of deep spinal muscles and to prevent the unloading-induced tonic-to-phasic shift of lumbo-pelvic muscle activation as it has been reported in a previous study after prolonged bed rest [6]. A biomechanical model of the FRED was developed and modified for a potential use in microgravity. It was hypothesized that both the modelled application of the modified device and the terrestrial model would lead to similar muscle recruitment patterns of lumbo-pelvic muscles.

Methods

In the following section, the acquisition of the experimental data for model validation, the model definition as well as its output are described. Furthermore the premises under which the model is validated and compared to its microgravity adaptation.

Experimental data

Intramuscular electromyographic data for the validation of the model were acquired in a previous study with the exercise device testing nine healthy male participants. This study was conducted at the Centre of Clinical Research Excellence in Spinal Pain, Injury and Health of the University of Queensland in Brisbane, Australia (manuscript submitted). The setup comprised intramuscular bipolar fine-wire electrodes (two Teflon-coated 75µm stainless-steel wires with 1 mm insulation removed from the ends, bent back to form hooks at 2 and 3 mm length, threaded into a hypodermic 0.50 x 70 or 0.50 x 32 mm-needle) which were inserted into the right side of the trunk under ultrasound guidance (Aixplorer, Supersonic Imagine, Aix-en-Provence, France).

The intramuscular EMG electrodes were placed in the following positions.

- Transversus abdominis (TrA): Between the anterior superior iliac crest and the lowest rib;
- Obliquus externus (OE): Between the anterior superior iliac crest and the lowest rib;
- Obliquus internus (OI): Between the anterior superior iliac crest and the lowest rib;
- Deep lumbar multifidus (LM): Between L4/L5, 30 mm laterally to spinous processes until the needle reached the most medial part of the L4 lamina
- Erector spinae (ES): At L2, 40 mm lateral to the spinous process

The participants were asked to perform exercise on FRED with a stable, upright trunk and were advised to maintain a constant angular velocity of 0.42 revolutions per second under three different amplitude settings. Visual feedback indicating the current exercise speed was provided. The smallest amplitude setting was used for the model validation.

The recorded data were amplified (factor 2000) and band-pass filtered in a range of 20 to 1000 Hz during acquisition. Data were recorded with a sampling frequency of 2000 Hz using a data acquisition system (Neurolog, Digitimer, Welwyn Garden, UK) and Spike 2 software (Cambridge Electronic Design, Cambridge, UK). Data were then digitally processed using Matlab (Version 2014a, Mathworks, Natick, MA, USA), and EMG signals were high-pass filtered (fine wire: 50Hz; surface: 30Hz), full-wave rectified and smoothed using a moving average filter with a time constant of 100 ms. A time window of 30 seconds was selected, and segments of EMG envelopes corresponding to individual movement cycles were time-normalized to 512 samples. For each participant, EMG signals corresponding to one specific muscle group were averaged and signals were amplitude-normalized according to the peak amplitude over all cycles. The muscle activation curves were averaged over all cycles. For the comparison between experimental data and simulation, muscle activities were averaged over all participants. The calculated standard deviation corresponds to deviations between the participants.

Model definition

The biomechanical model used in this study was developed with the Anybody Modeling System^{TM} (version 6.0.3, AnyBody Technology A/S, Aalborg, Denmark). The design of the model is based on the newest terrestrial prototype of the FRED ($FRED_{oneG}$) and it was then adapted for a microgravity environment ($FRED_{zeroG}$) using the human body model (HBM) in the software environment. The HBM, as provided by AnyBody Technology, consists of several body groups which are designed individually for specific purposes. The resulting HBM, the AAUHuman model, was constructed from the ShoulderArm Model, the Cervical Spine Model, the LegTLEM Model and the Lumbar Spine Model. The latter one, being most important for this study, consisted of 5 vertebra, 188 muscle fascicles, spherical joints with 3 degrees of freedom as well as a model for intra-abdominal pressure. It was designed according to data taken from previous studies [9, 15].

The Functional Re-adaptive Exercise Device was constructed as a mechanical part of the model and consisted of eight parts for the $FRED_{oneG}$ model. These include the wheel (a), two horizontal (b) and two vertical bars (c), a bracket (d), which mounts the two vertical bars to one another, as well as two connecting segments (e) that link the back sides of the left and right horizontal bars to specified nodes on the wheel (Figure 2). The full assembly was grounded in two nodes to fix points in the global reference frame, one in the centre of the wheel (a) and the other one in the centre of the bracket (d). All components were connected via hinges, allowing only revolutions about one specific axis, resulting in one degree of freedom per joint. In order to avoid redundancy in the model, the connecting segments which link the horizontal bars (b) to the wheel were attached to it by constraining all six degrees of freedom.

The ideal movement on the FRED comprises a stable trunk and an even, quasi-elliptical movement of the legs, with legs being near to full extension at the lowest point of the elliptical foot motion. The upper body should be upright (0 deg) while arms should be held to the side of the trunk (i.e. not used to provide any support). Rotational movements of the spine should be avoided and the exercise velocity should not exceed one revolution per second. In order to achieve this, the HBM was connected to the FRED and to the global Anybody™ reference frame using three reference points on the skeleton (Figure 2): one reference point at each foot (f) and one at the sacrum (g). Movements of the reference points (f) were restricted to only one degree of freedom (DoF) allowing lifting of the feet along the local y-axes of the horizontal bar segments, as shown in Figure 2. Reference point (g) was constrained in all six degrees of freedom at a defined level and orientation relative to the global reference frame, to ensure a stable trunk and a position with almost extended legs on the lowest point of the elliptical foot motion. Reaction forces were only considered for the feet (reference points f). Masses and moments of inertia of the FRED components were not taken into account in the model. The motion was driven by a constant angular velocity of 0.42 revolutions per second matching the settings of the intramuscular study. The forces and moments required for the desired motion were generated by the HBM

The principle for the modifications to adapt the $FRED_{oneG}$ model to the microgravity model was adapted from the current treadmill set up that is currently used on-board the International Space Station (ISS). Treadmill users on the ISS are required to wear a harness that consists of elastic straps to generate loading in the y axis [13]. The forces that, in reality, are applied by elastic straps linked to the harness system, were simulated by vertical forces attached to the HBM. For the $FRED_{zeroG}$ model, the gravitational vector was removed and instead, two vertical forces (F1, F2) were applied to the shoulder reference points, 17 centimetres apart from the spine with:

$$F_1 = F_2 = \frac{1}{2}\, m \cdot\ g \cdot m_{UBrel}$$

where m is the body mass, g is the acceleration due to gravity on the Earth's surface (9.81 m/s^2) and m_{UBrel} is the relation of the upper body mass as a percentage of the total body mass, which, according to reference values presented by Salvendy [29], is 44.68%. Thus, the resulting loading pattern of the spine should simulate earth-like conditions for the upper trunk.

Model output

The AnyBody software calculates muscle forces during motion using inverse dynamics [26]. As large parts of the muscular system of the human body are developed redundantly, the computation of muscle forces would lead to infinitely many solutions of the equations without the implementation of restrictions, making it an indeterminate system. An optimization function is therefore introduced which follows the principle that the most efficient combination of muscle actions across a certain joint will be taken.

For some muscles (e.g. multifidus), the muscle is constructed as a number of individual muscle strings. All strings which belong to one muscle group were averaged in order to obtain a model output for the entire muscle to enable comparisons with measured electromyographic data.

The $FRED_{oneG}$ and $FRED_{zeroG}$ models were both simulated for one complete movement cycle over a period of 2.381 seconds with the angular velocity of the wheel set to 0.42 revolutions per second matching the target speed of the intramuscular EMG study. To generate a modelling output which is comparable to EMG data, the AnyBody Modelling System provides muscle activation based on a simple muscle model relating a muscle's force linearly to its activation and normalizing it to its maximum exertion force. This approach shows reasonable results for small joint angle changes and moderate contraction velocities.

Model validation

For the muscles TrA, OE, OI, LM and ES intramuscular EMG data and the $FRED_{oneG}$ model output were compared to determine the validity of the model. The model was deemed valid if it matched the measured data for the following two criteria:

- Is the muscle recruited during the entire movement? – Positive if normalized EMG and modelled activation levels are greater than 0% during the entire cycle.
- Are the derivatives of the muscle activation curves similar? – Positive if the difference between the derivative of the modelled and the measured muscle activities is < 5%.

As stated above, in FRED exercise a tonic and continuous muscle activation is desired. Therefore one of the criteria to determine the credibility of the model is to compare the continuity of the muscle activation.

Furthermore it is important to distinguish between a more tonic-fashioned activation (continuous activation with minimal amplitude changes) and phasic-fashioned activation (higher amplitude changes).Thus, the derivative of the curve is calculated. The derivative of the EMG data gives a measure of the tonicity of the muscle recruitment. A tonic activation will have a derivative that is closer to 0% than a more phasic muscle recruitment pattern. The validation was assumed to be successful if the error between the two derivatives was less than 5%. It is important to note that, due to the different normalisation methods used in the intramuscular EMG measurements and the modelled muscle activations, it was not appropriate to compare the normalised muscle contraction amplitudes as part of the model validation.

$FRED_{oneG}$ versus $FRED_{zeroG}$

In order to investigate the potential for the FRED to be used as an in-flight countermeasure to address the negative effects of microgravity on the lumbo-pelvic muscles, the modelled muscle outputs for TrA, OE, OI, LM and ES were directly compared. Both models were driven using a constant angular velocity of 0.42 revolutions per second of the wheel, relative to the reference frame, without applied reaction forces to ensure consistency of movements between the models.

The extracted data were then compared assessing the type of contraction (tonic, phasic) and the magnitude of the modelled muscles as a proportion of the peak muscle activation in the cycle.

Results

The results as obtained from model validation and the comparison between $FRED_{oneG}$ and $FRED_{zeroG}$ are presented in the following section.

Model validation

The results obtained from the model validation are depicted in Figures 3-7. Each figure contains one graph showing the muscle activations (a) and the corresponding derivatives (b).

Figure 3 shows the behaviour of TrA. It can be seen that the muscle was continuously active for both the simulation and the intramuscular EMG measurements. The modelled activation showed a mostly tonic contraction of around 90% with two minor phasic peaks at around 45% and 95% of the cycle. The amplitude of the measured TrA EMG muscle activation was approximately 30% and did not show phasic activation bursts. The standard deviation between the participants for TrA was approximately 5% on average throughout the cycle. For both the intramuscular EMG data and the modelled data it was found that the average derivative is 0%. The difference between the two curves was smaller than 0.5%.

The modelled and measured activities of OE are depicted Figure 4. The muscle activation as calculated with the model showed a constant activation of around 60% with a major peak at around 90% and a minor peak at around 45% of the cycle. The measured muscle activation revealed a tonic contraction at around 45% with minor fluctuations indicated by the standard deviation of approximately 5%. The derivatives of EMG signal and model output were 0% throughout most of the cycle but with minor deviations at around 45% and major deviations at 0% and 90% for the modelled data. The maximal difference between the derivatives was found to be 2.5%.

The results of the $FRED_{oneG}$ model and the measured activation of OI are shown in Figure 5. The muscle activations of both $FRED_{oneG}$ and intramuscular EMG were around 42%. While $FRED_{oneG}$ showed a tonic-fashioned behaviour with only one major peak at around 90% of the cycle, the measured activation remained mostly constant with small standard deviations of around 2.5%. The derivatives of both EMG signal and model output were mostly constant around 0% with peaks of -2% and 2.5% at 0% and 85% of the cycle for the model.

The EMG data and $FRED_{oneG}$ muscle activation obtained for LM are presented in Figure 6. The model revealed a tonic contraction with minor peaks at around 45% and 90% of the cycle. Although the measured activation showed a tonic activation pattern, its amplitude was lower with an average of about 25%. The standard deviation of the EMG data was around 5%. Both, the derivatives of the EMG signal and the model output were found to be around 0% with fluctuations of the modelled data at around 40% and 90% of the cycle. The maximal difference between the two derivatives was around 1%.

Erector Spinae showed an activation of about 80% in the simulation and of 25% during the EMG measurement. The modelled muscle activation as presented in Figure 7 showed two peaks at 45% and 95% of the cycle. The measured EMG activation was nearly constant with a standard deviation of about 7.5%. The curves representing the corresponding derivatives of EMG signal and model output show for both cases minor fluctuations around an average of 0% with a maximum difference of around 0.5%.

Table 1 illustrates the overall outcome of the validation. It shows that both of the two validation criteria were met for all muscles. All muscles were recruited for 100% of the movement cycle in both the measures and modelled data. There was also less than 5% difference in the derivatives of the muscle activity data between measured and modelled data. This, therefore, confirmed the validity of the model.

$FRED_{oneG}$ vs$FRED_{zeroG}$

The comparison between the $FRED_{oneG}$ and the adapted micro-G $FRED_{zeroG}$ model is presented in Figures 8a) to e).

Figure 8a) shows the activation of TrA of both the $FRED_{oneG}$ and $FRED_{zeroG}$ models. $FRED_{zeroG}$ revealed a constant activation of 100% while minor phasic peaks were observed for $FRED_{oneG}$ at 45% and 90% of the cycle, which fluctuated around 90%.

The comparison in regards to OE is shown in Figure 8b). It can be observed that OE was activated in a more phasic fashion in $FRED_{zeroG}$ than in $FRED_{oneG}$, both with peaks at around 50% and 90% of the cycle. The average muscle activation of OE was lower in $FRED_{zeroG}$.

Comparing the OI activation between both models showed a similar image. As depicted in Figure 8c), both activities were high in the beginning and end of the cycle, with a major peak at 90% and a constant activation between around 30% and 80%.

Figure 8d) visualizes the activities of LM in both simulations. In both cases LM was activated in a highly tonic fashion with an average of about 95% for $FRED_{zeroG}$ and 90% for $FRED_{oneG}$. $FRED_{oneG}$, however, shows slightly more phasic behaviour.

The comparison of ES for both models, which is shown in Figure 8e), indicated highly tonic activation for $FRED_{oneG}$ and $FRED_{zeroG}$. The average activation in $FRED_{zeroG}$ was slightly lower with a minimum at around 50% of the cycle.

Discussion

The main finding of the present investigation was that the deep spinal muscles are activated during the modelled $\mathrm{FRED}_{\mathrm{zeroG}}$- exercise, supporting the idea that the principle of FRED exercise as it is used on Earth can be transferred to microgravity conditions, leading to comparable muscle recruitment patterns of lumbo-pelvic muscles. Moreover, the applied model was shown to be valid when comparing real muscle activities and the activation derived from the terrestrial FRED model. The present study suggests that when performed in a microgravity environment, FRED exercise has the potential to directly address spaceflight specific muscular deficits [12, 25]**.** Consequently, on-board FRED-exercise could be useful to maintain astronaut health in space and to reduce the increased risk of back injuries upon their return to Earth [3, 23].

Model validation and $\mathrm{FRED}_{\mathrm{oneG}}$

For all muscle groups the validation criteria were fulfilled, which is shown in Table 1. All muscle groups behave in a desired, mostly constant fashion, and the activity patterns can be considered similar to the experimental data. Notable differences in the amplitudes of the activities can be observed. This, however, could be caused by reasons described under *Limitations of the study.*

Previous terrestrial studies with the FRED reported promising results, indicating that it could be a valid tool to be used in people with deficits of deep spinal and 'anti-gravity' muscles [7, 10, 14]. In the present study, the computed model activities of lumbo-pelvic muscles during FRED exercise are in line with previous physiological studies [10, 14] and reinforce the findings that the combination of lower-limb movement and weight-bearing in a constant and slow motion shows great potential to activate these muscles [7, 10, 14]. The results from the $\mathrm{FRED}_{\mathrm{oneG}}$ model indicated that all of the investigated lumbo-pelvic muscles are constantly activated throughout the whole exercise cycle. The activity levels of the relevant muscles as calculated for $\mathrm{FRED}_{\mathrm{oneG}}$ show minor phasic behaviour for OI and ES with a majorly tonic activation of OE. Importantly, the deep spinal muscles TrA and LM are both constantly activated during the simulated exercise.

The modelled data of the terrestrial FRED reinforce the findings of previous FRED investigations [7, 10, 14] and support the idea that FRED exercise could be used in astronaut rehabilitation and rehabilitation of people on Earth with deficits in spinal stability and postural control, such as people suffering from low back pain [4, 20].

FRED$_{zeroG}$

The microgravity model of the FRED suggests that TrA and LM are more active compared to the terrestrial FRED model. The modelled FRED$_{zeroG}$ activation pattern of the lumbo-pelvic muscles is very promising as it shows that FRED$_{zeroG}$-exercise may be able to specifically address the deep spinal muscles. Although the trunk flexor muscles are also activated to some degree, the modelled trunk muscle activation patterns show great potential to directly counteract unloading-induced muscular imbalances and unfavourable motor control strategies of lumbo-pelvic muscles as reported after space flight and bed rest [5, 6, 25]. As current on-board countermeasures do not specifically address deep spinal muscles, it could be argued here that in-flight exercise with FRED may be able to prevent the microgravity-induced atrophy of these muscles and eventually minimize the increased risk of herniated discs, prevent muscular imbalances [5] and counteract disc degradation [4, 5] associated with this atrophy [23].

Future considerations

As a first step, the long term benefits of the terrestrial version of the FRED should be investigated. The population of choice to do that would be bed rest deconditioned individuals as they experience very similar adaptations to the lumbo-pelvic region as astronauts do [3, 5]. Additionally, further intramuscular fine wire EMG studies could be used to study trunk muscle activation on the so-called vertical treadmill or during treadmill walking/running in a parabolic flight. This would allow a direct assessment of the specific effects that harnesses and elastic bands have on trunk muscle activation patterns. Production of a ‘zero G’- prototype of the FRED should be considered once these studies have been completed and provided that the long term effects of the FRED would reinforce the use of it as a countermeasure. Also, from a theoretical perspective the application of two symmetric, vertical forces on the shoulders of the exercising subject shows good results in providing load. In reality, however, the idealized punctuated forces that are used in the model would need to be distributed over a larger surface area than just the shoulders to avoid high, localized and uncomfortable ‘hot spots’. Padding or cushioning between the elastic straps and the body should be considered to alleviate this condition.

Limitations of the study

With regards to the design of the biomechanical FRED models, the idealized posture and motion of the HBM might have an influence on the outcomes of the study. It is possible that in reality, the exercising subject might have coordination deficits whilst performing the exercise, being unable to maintain a perfectly upright posture. The evenness of the motion is also idealized in the biomechanical model and could, in reality, be of importance for the muscle activation patterns. Two other factors that are not considered in this model are reflexes and proprioception which are known to have a great impact on muscle activation [2].

The absence of proprioceptive behaviour in the model might also have an impact on the overall muscle recruitment patterns. The optimization criterion determining each muscles' contribution to the overall motion favours synergy between them. This results in an overall continuous model, while proprioceptive reflexes induce rapid changes in muscle activations.

However, the biomechanical model of FRED exercise can be regarded as a first step in the investigation process of a promising in-flight exercise countermeasure principle.

Conclusion

From a biomechanical model perspective, the results of this work show that the principle of terrestrial FRED exercise is transferable to a microgravity environment. The modelled muscle activation patterns during $FRED_{zeroG}$ exercise are promising and may be able to address astronaut-specific deficits. In-flight exercise on the FRED could potentially help to maintain astronaut back and core health in the long term. The findings of this work provide the first indication of how to improve the design of the device for this purpose. Further investigations are needed to acquire physiological *in vivo* data.

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Tables

Table 1. Model validation of transversus abdominis (TrA), external obliques (OE), internal obliques (OI), lumbar multifidus (LM) and erector spinae (ES) for intramuscular EMG and $\mathrm{FRED}_{\mathrm{oneG}}$ outputs.

Figure 1. The current prototype (Mark II) of the FRED with an exercising subject

Figure 2. The AnyBody model of the FRED

Figure 3. (a) Intramuscular electromyographic data in comparison to muscle activity as simulated using $FRED_{oneG}$ transversus abdominis (TrA) and (b) derivatives of intramuscular electromyographic data in comparison to derivatives of muscle activity as simulated using $FRED_{oneG}$ for TrA.

Figure 4. (a) Intramuscular electromyographic data in comparison to muscle activity as simulated using $FRED_{oneG}$ external obliques (OE) and (b) derivatives of intramuscular electromyographic data in comparison to derivatives of muscle activity as simulated using $FRED_{oneG}$ for OE.

Figure 5. (a) Intramuscular electromyographic data in comparison to muscle activity as simulated using $FRED_{oneG}$ internal obliques (OI) and (b) derivatives of intramuscular electromyographic data in comparison to derivatives of muscle activity as simulated using $FRED_{oneG}$ for OI.

Figure 6. (a) Intramuscular electromyographic data in comparison to muscle activity as simulated using $FRED_{oneG}$ lumbar multifidus (LM) and (b) derivatives of intramuscular electromyographic data in comparison to derivatives of muscle activity as simulated using $FRED_{oneG}$ for LM.

Figure 7. (a) Intramuscular electromyographic data in comparison to muscle activity as simulated using $FRED_{oneG}$ erector spinae (ES) and (b) derivatives of intramuscular electromyographic data in comparison to derivatives of muscle activity as simulated using $FRED_{oneG}$ for ES.

Figure 8a) – e). $FRED_{oneG}$, mucle activity in comparison to $FRED_{zeroG}$ muscle activity

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Table1

Muscle	**Muscle constantly active throughout the whole cycle?**		**Difference between derivatives less than 5%**		**Validation successful?**
	Intram. EMG	*FRED$_{oneG}$*	*Intram. EMG*	*FRED$_{oneG}$*	
TrA	Yes	Yes	Yes	Yes	Yes
OE	Yes	Yes	Yes	Yes	Yes
OI	Yes	Yes	Yes	Yes	Yes
LM	Yes	Yes	Yes	Yes	Yes
ES	Yes	Yes	Yes	Yes	Yes

Figure1

Figure2
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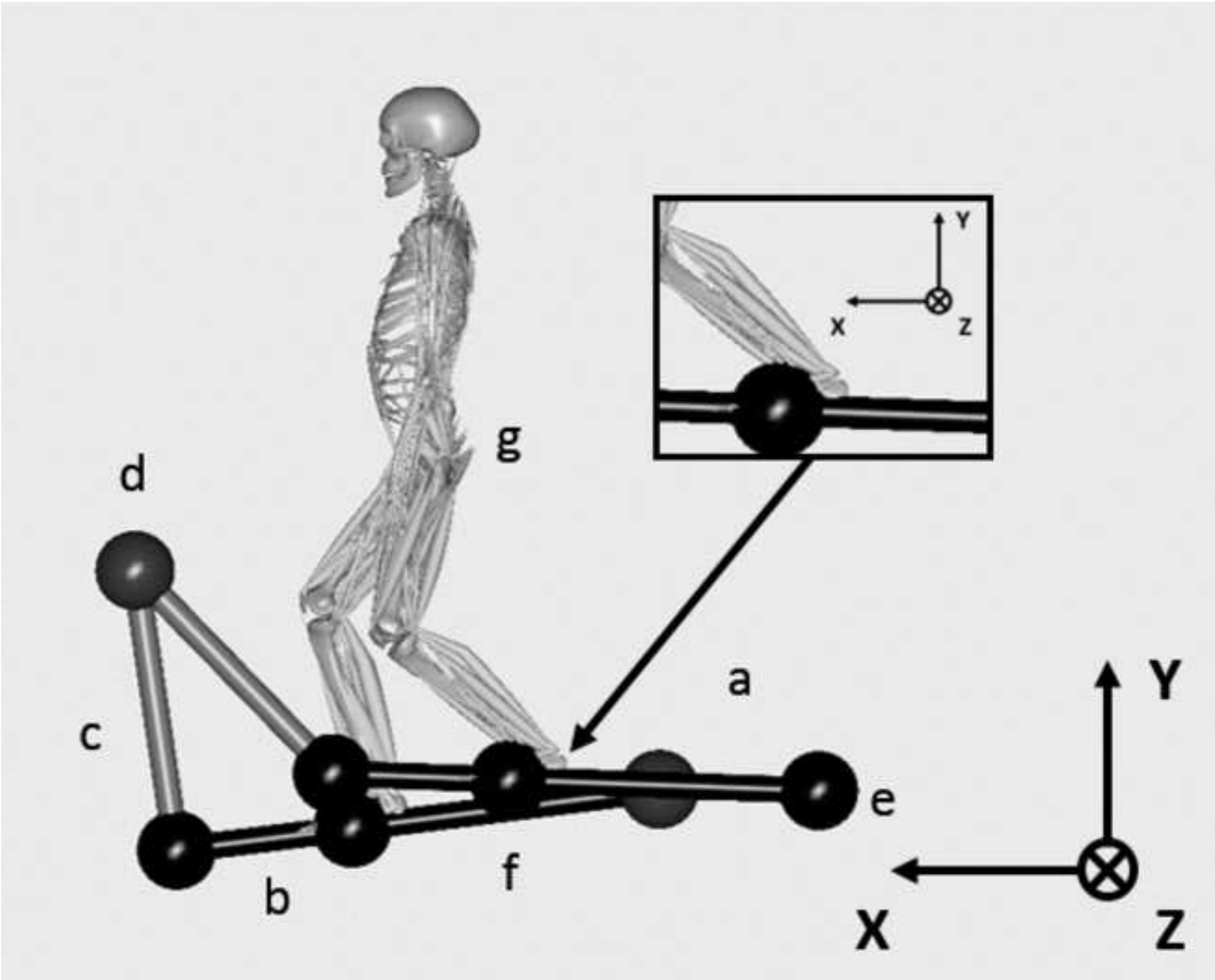

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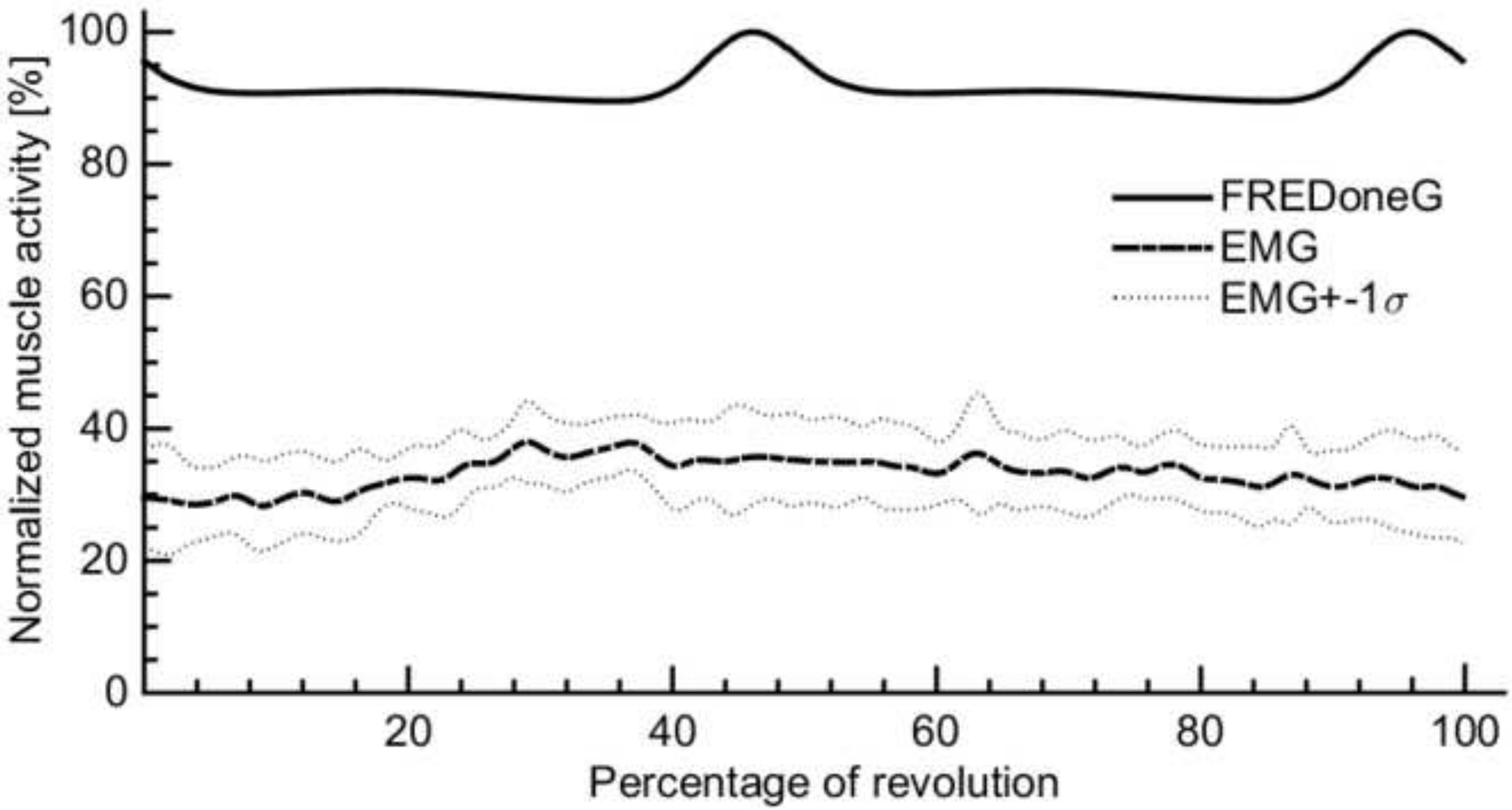

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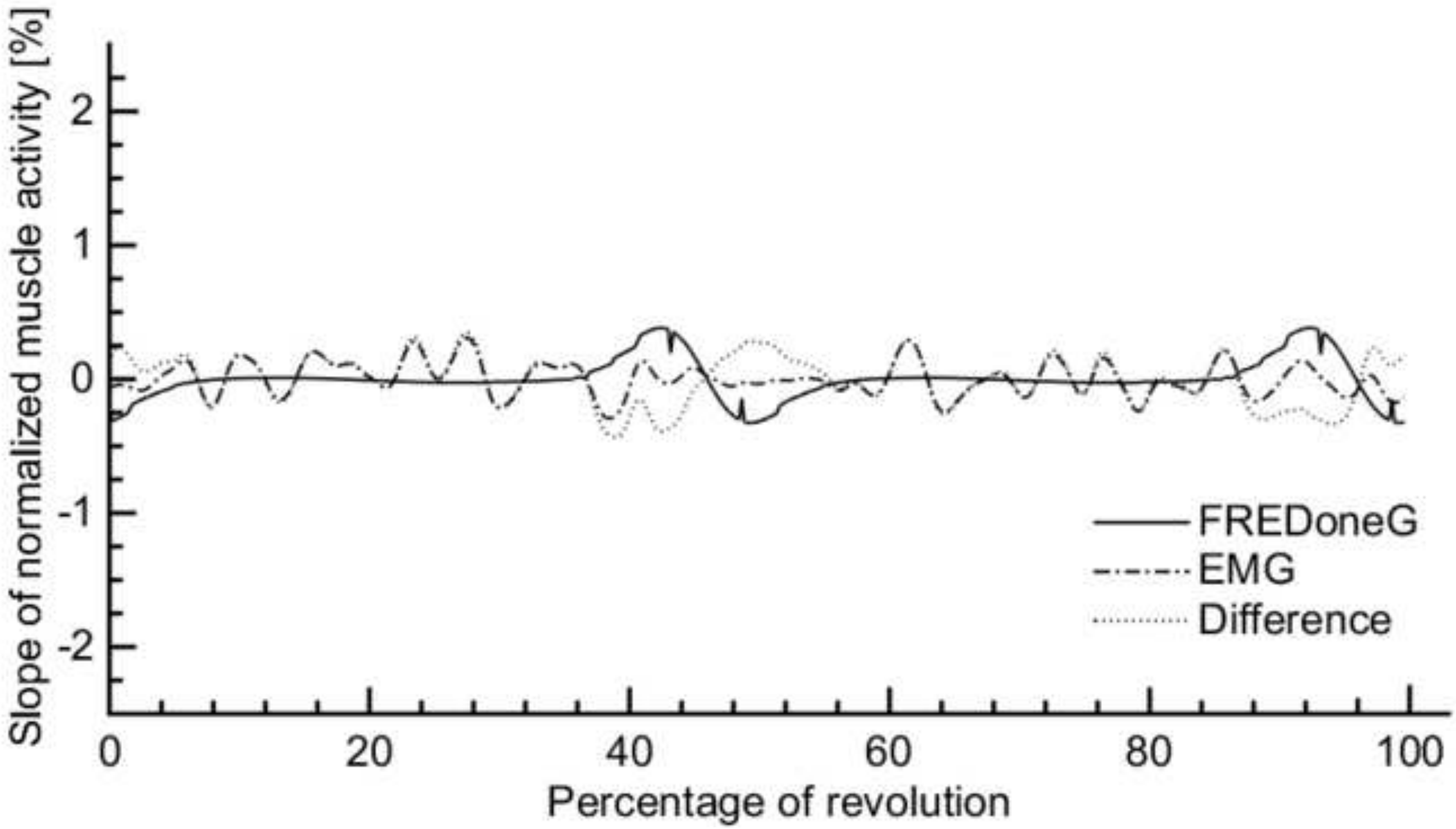

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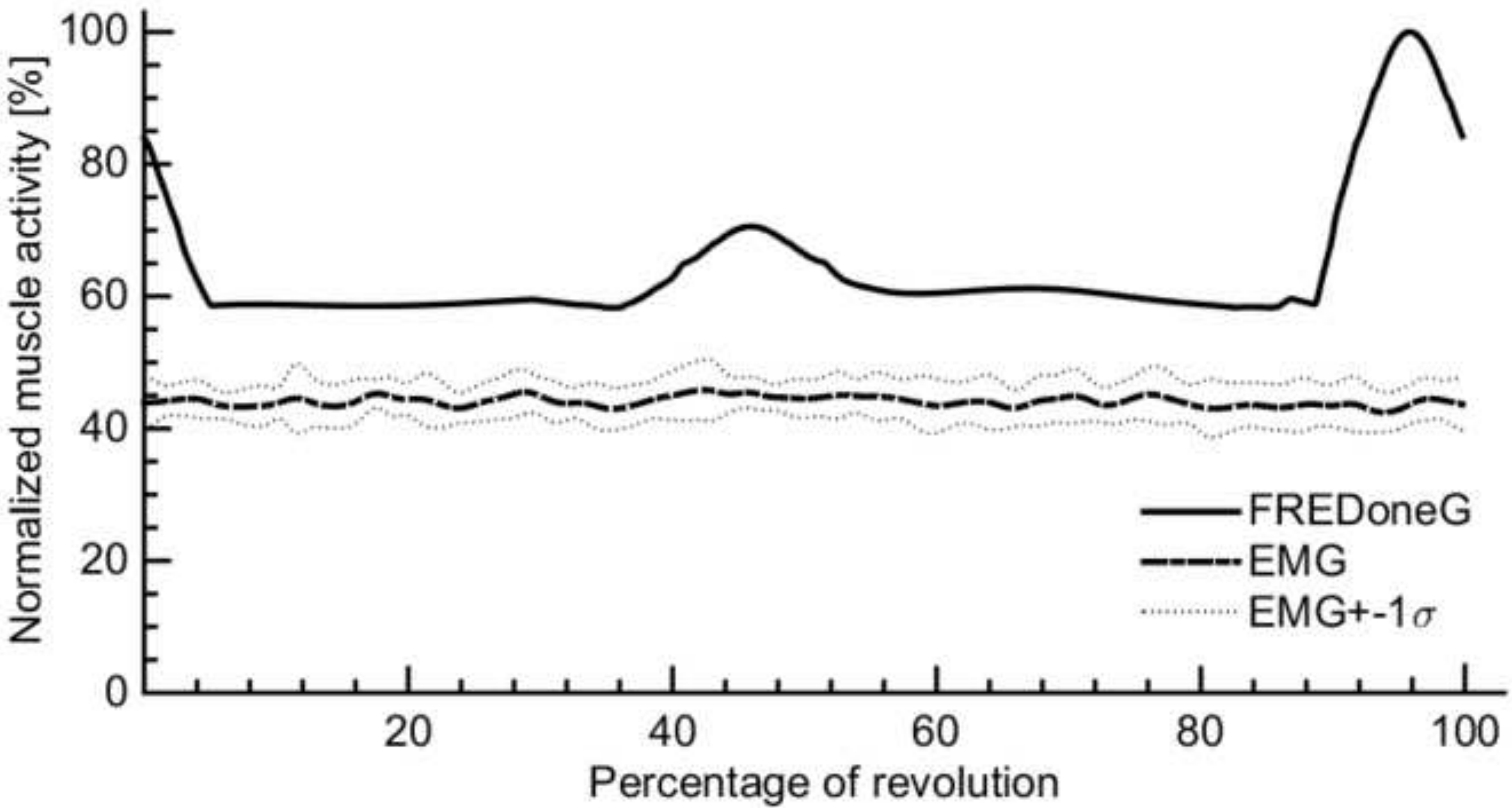

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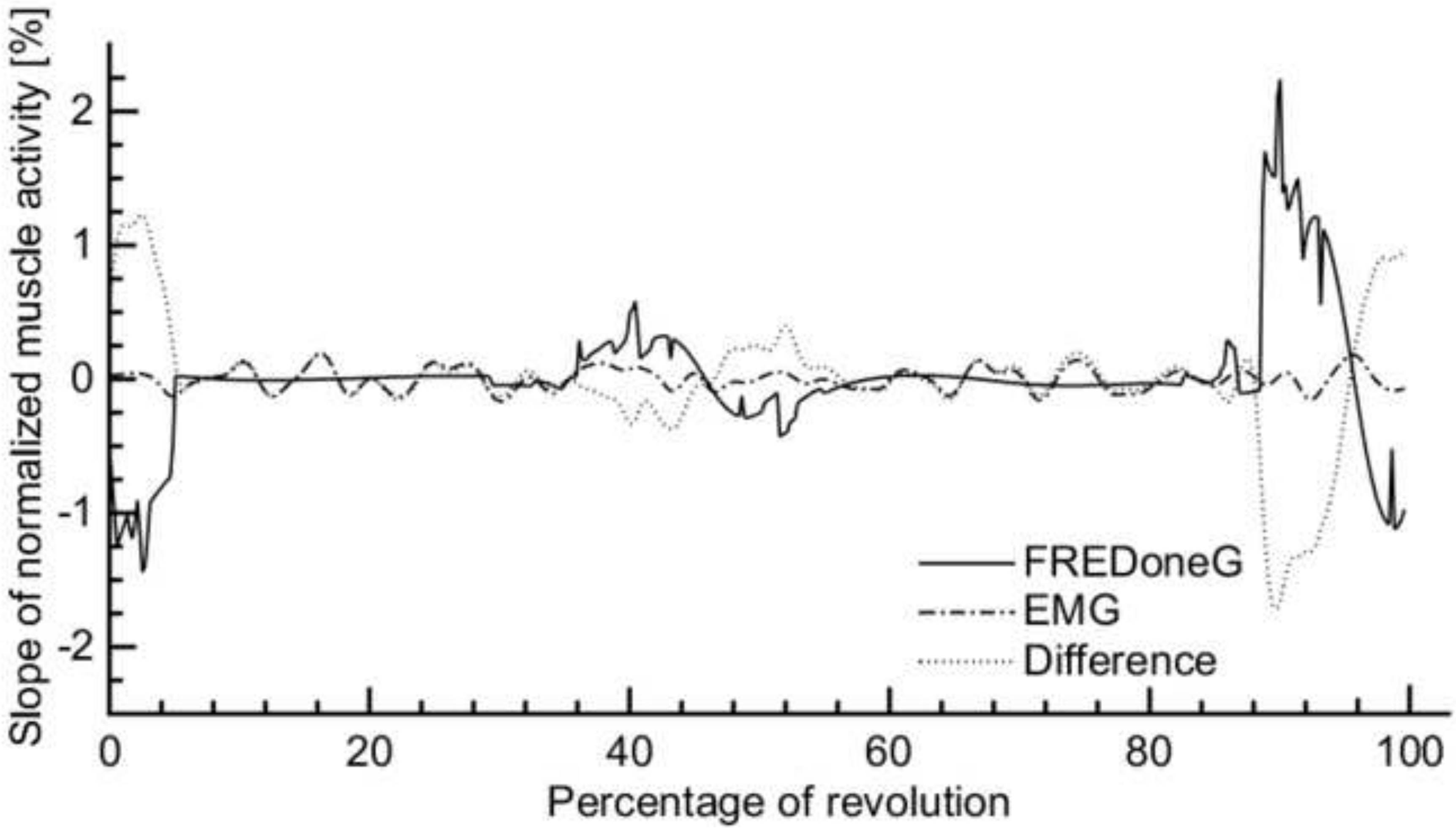

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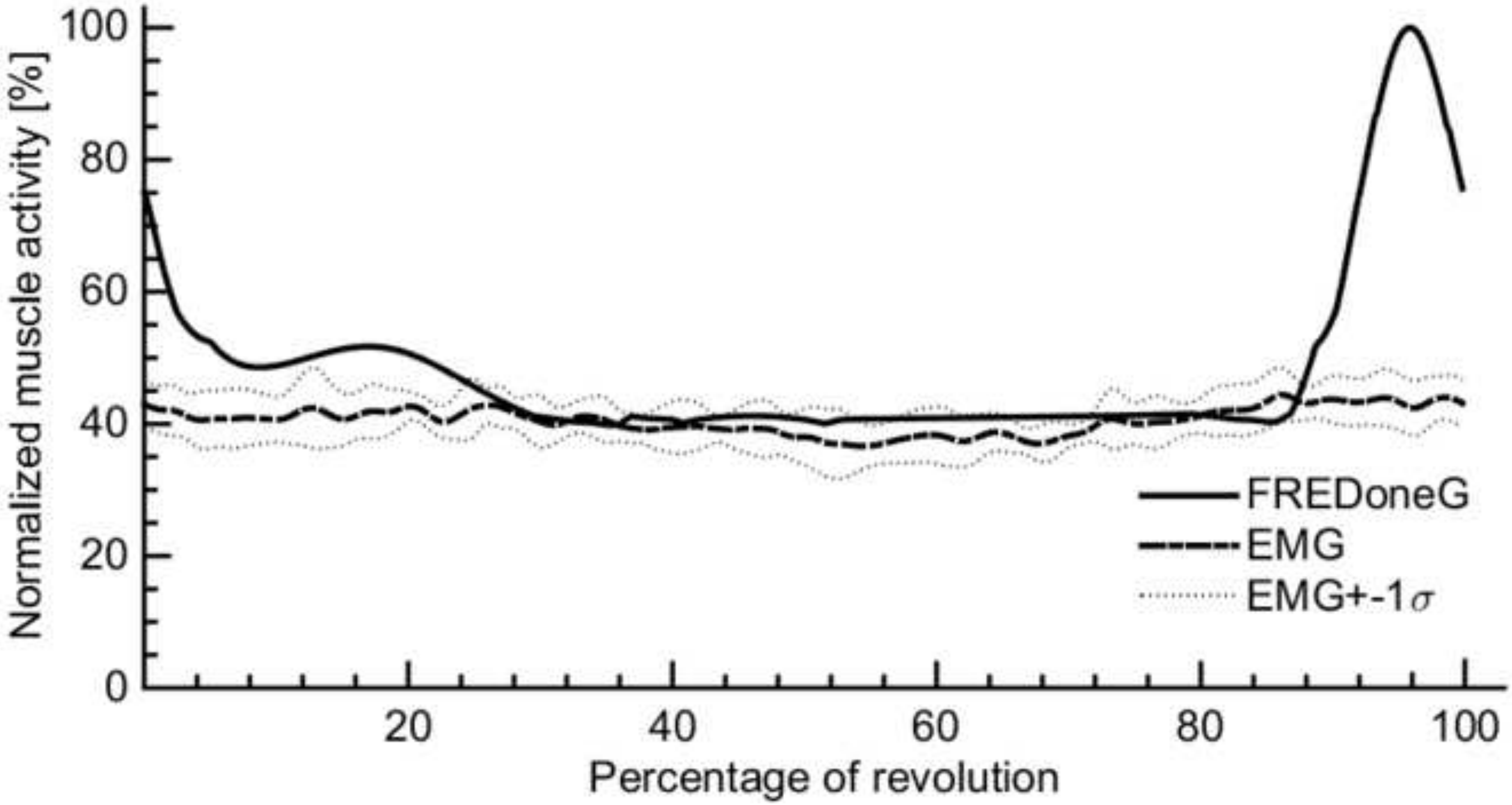

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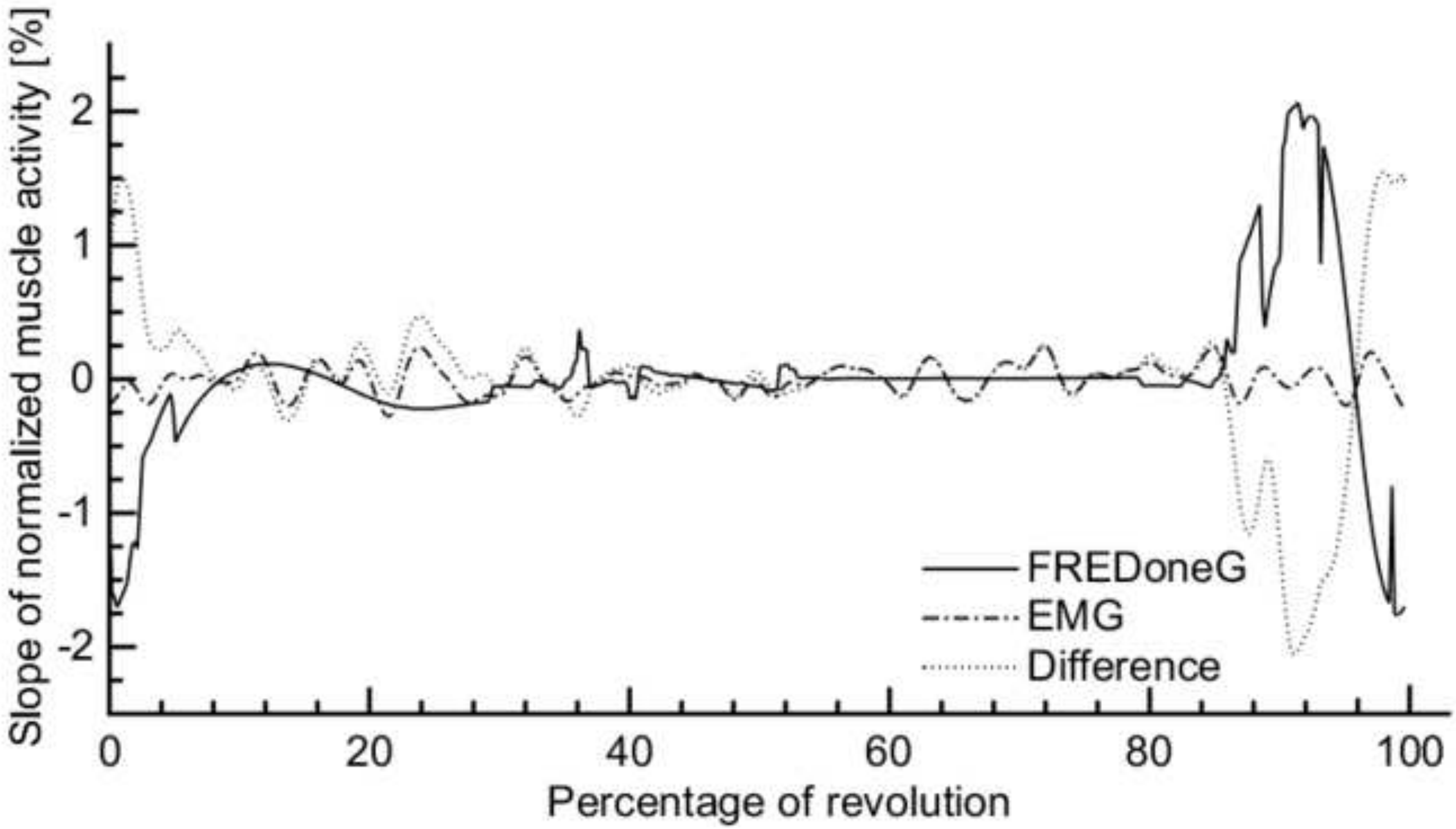

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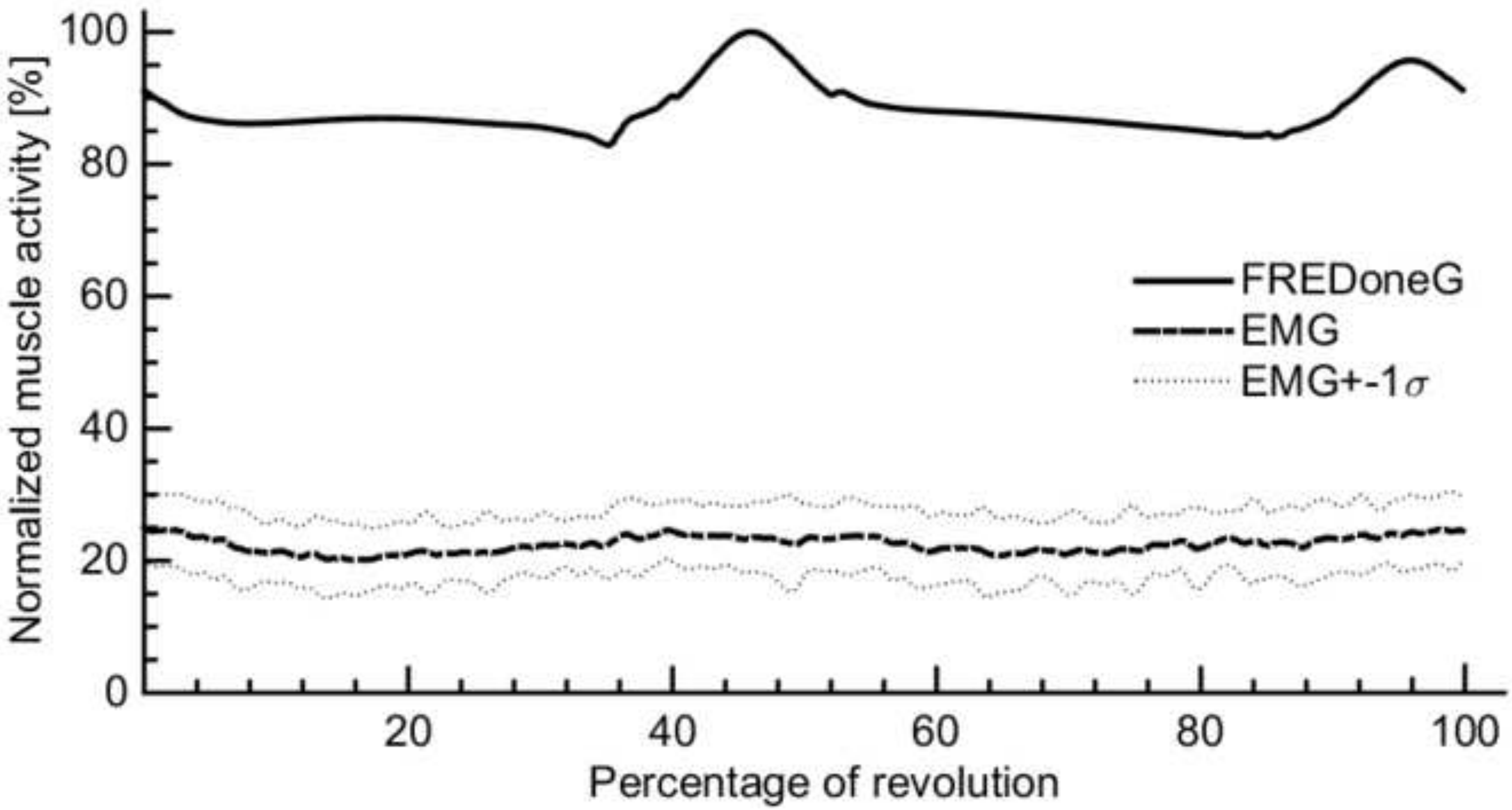

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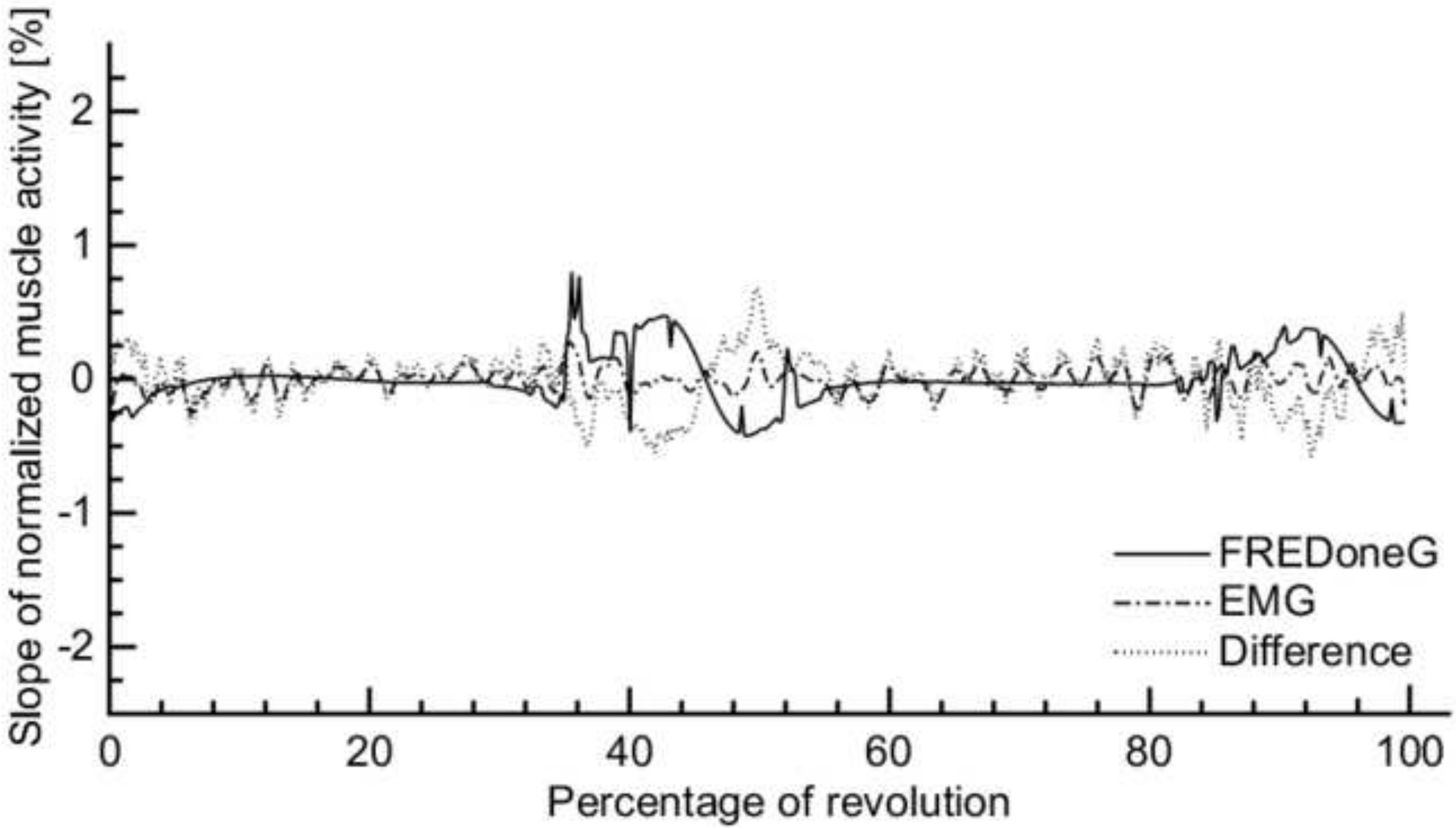

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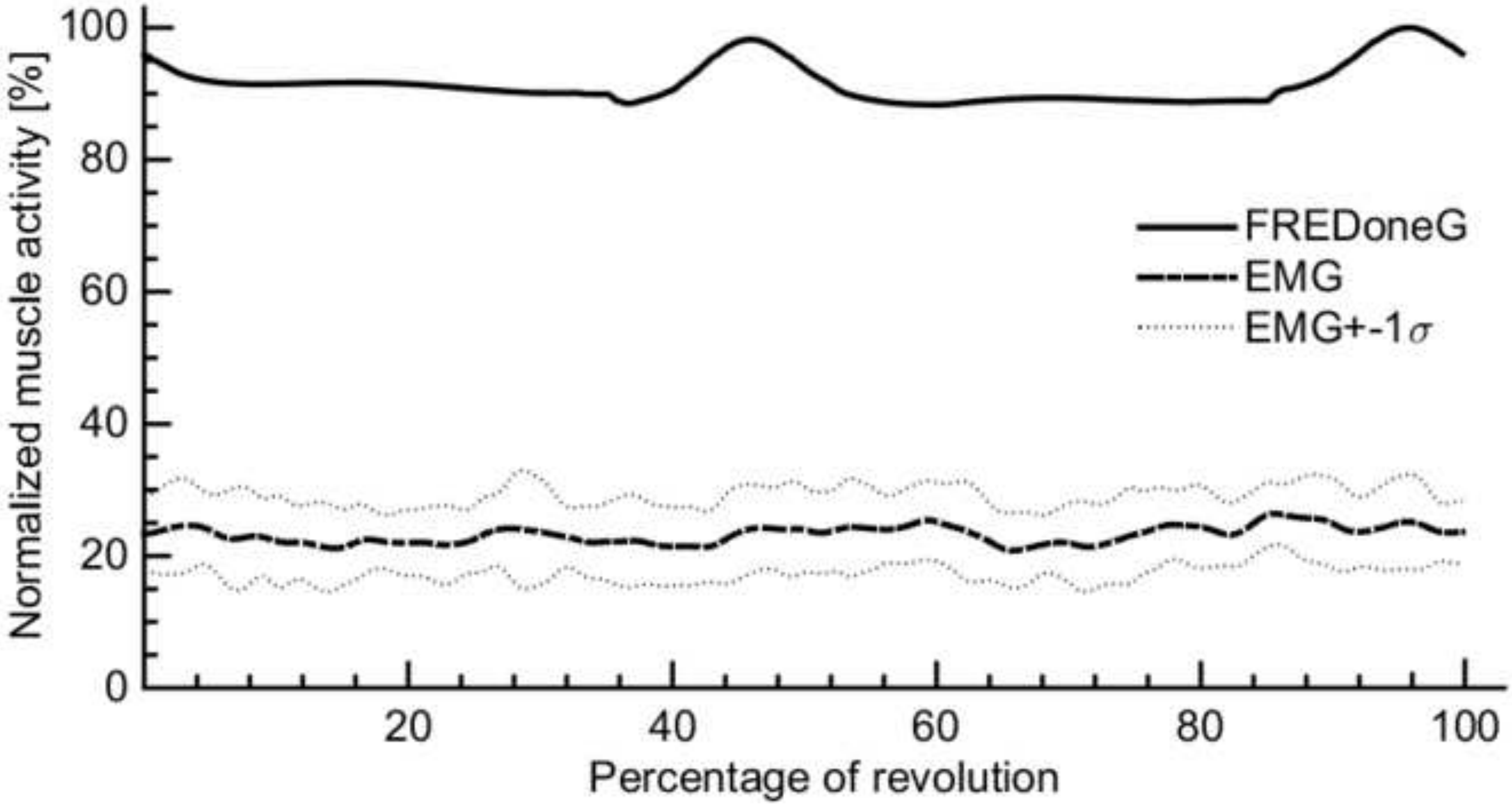

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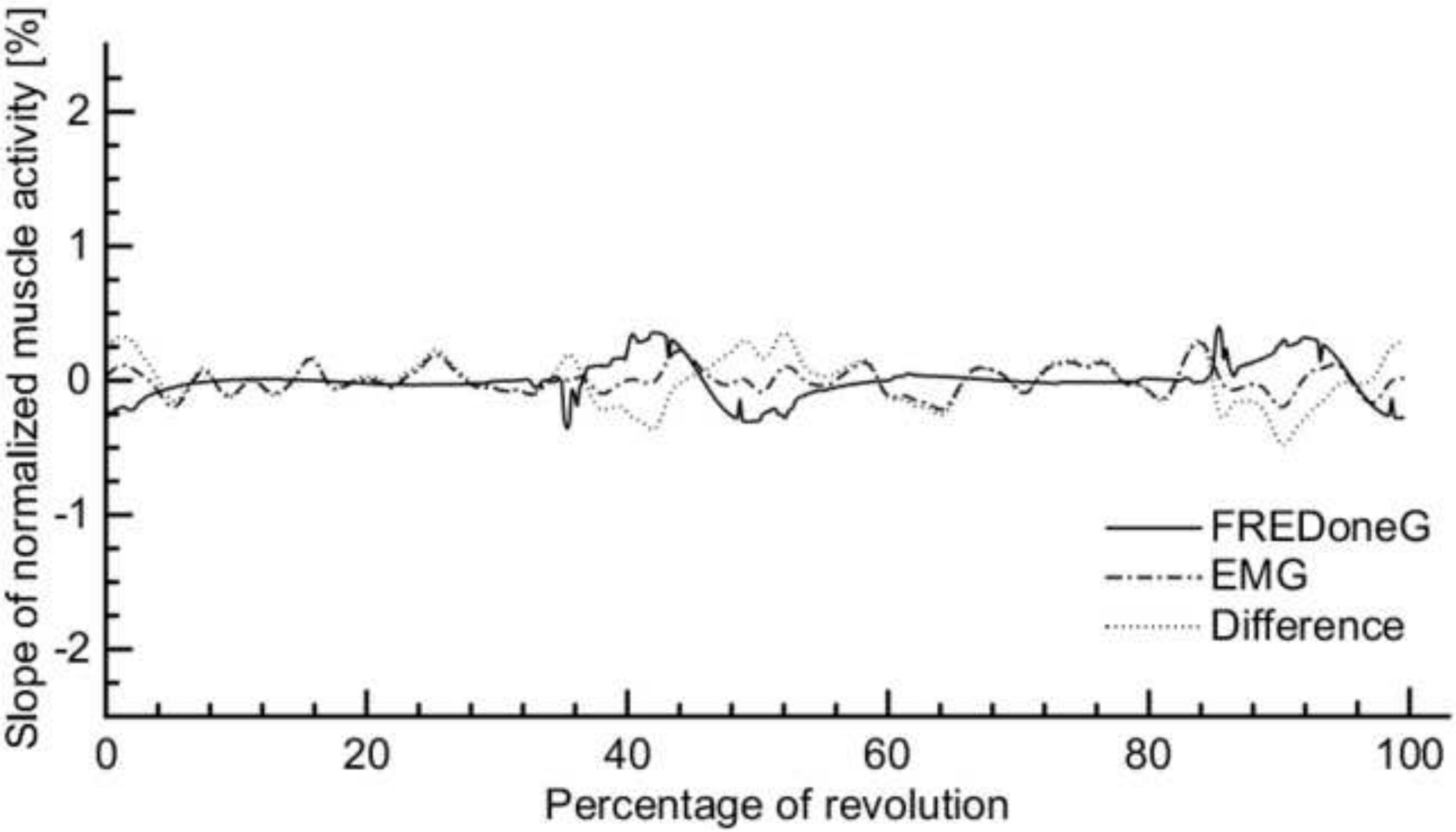

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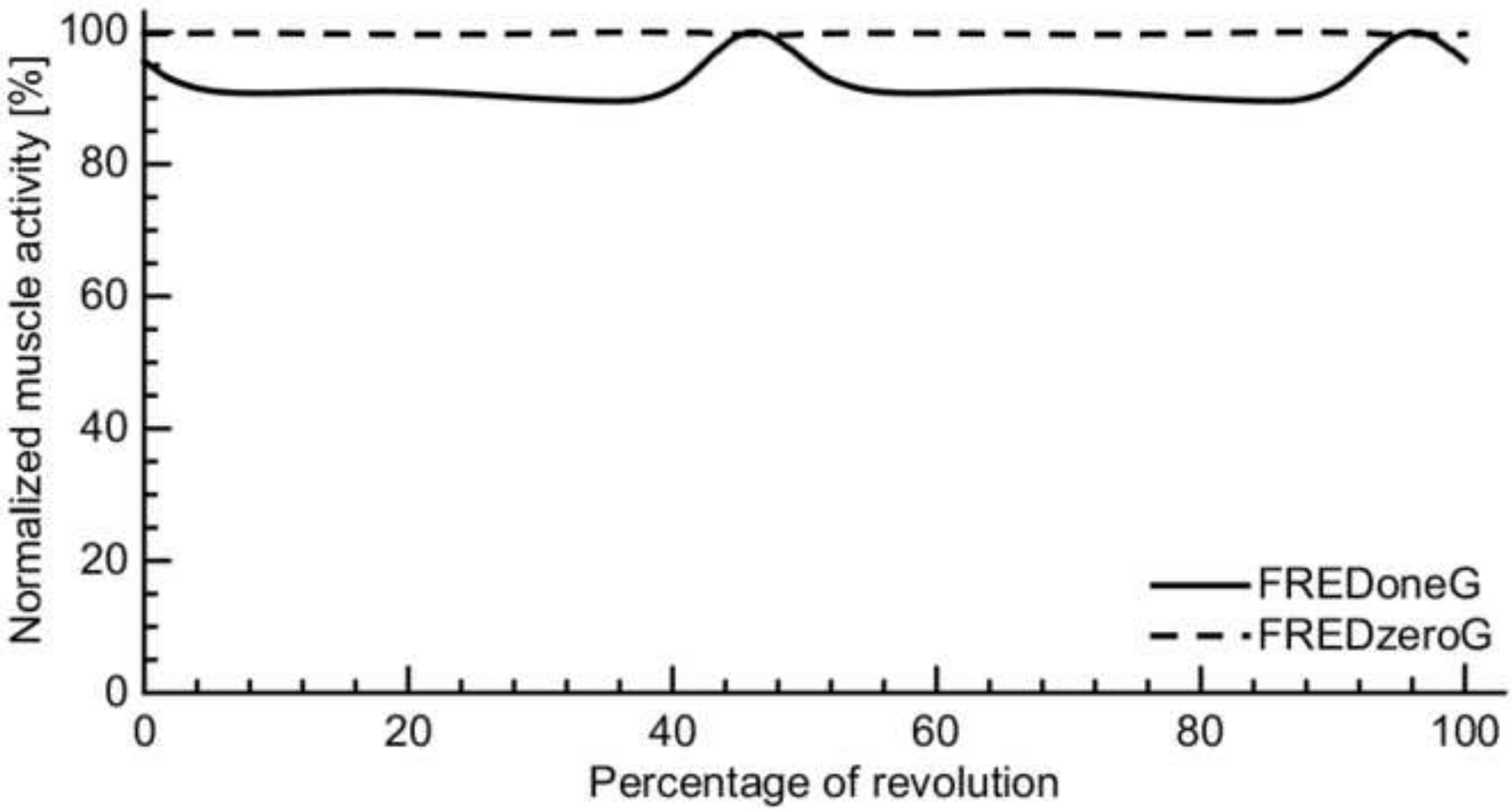

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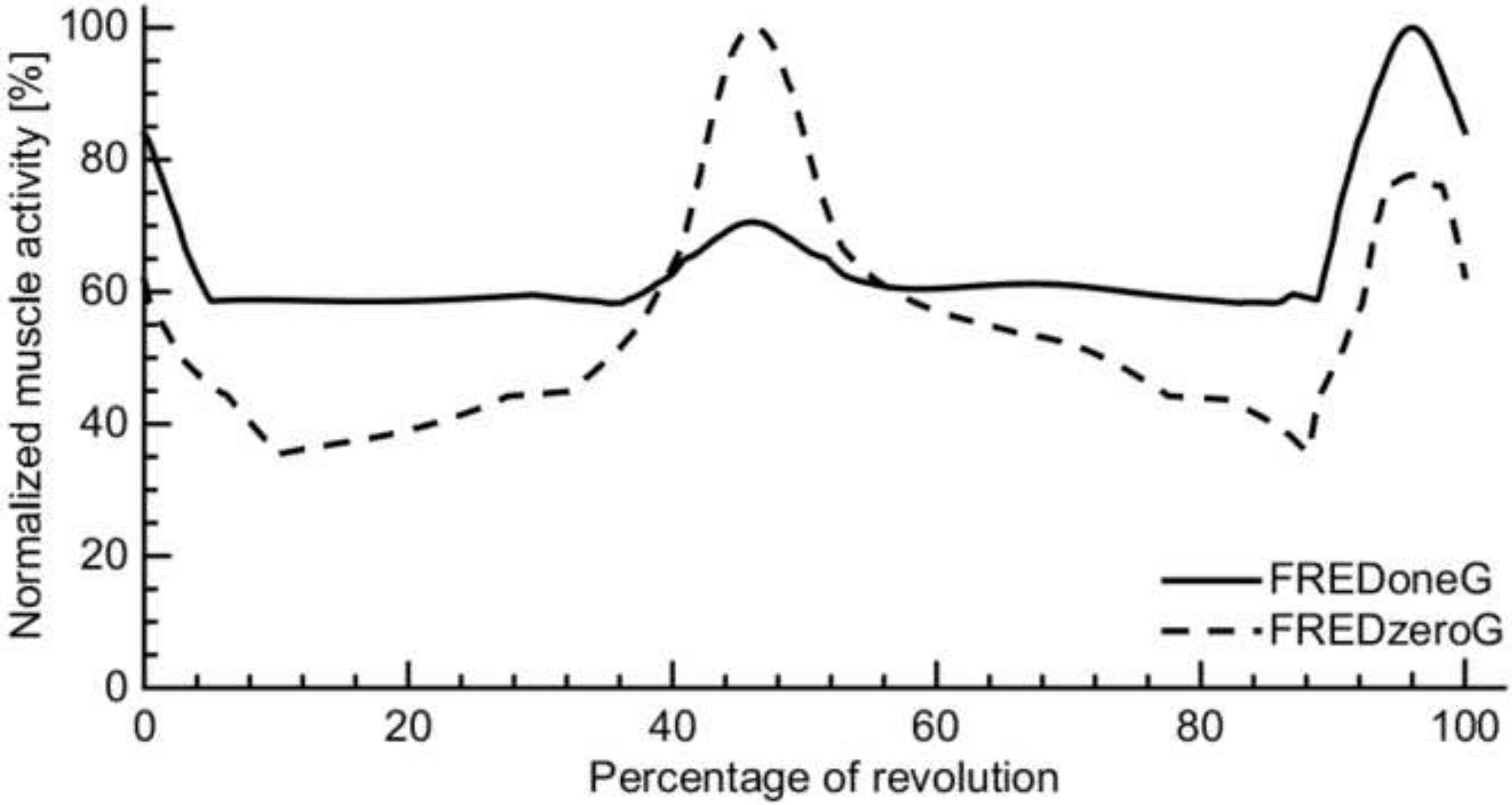

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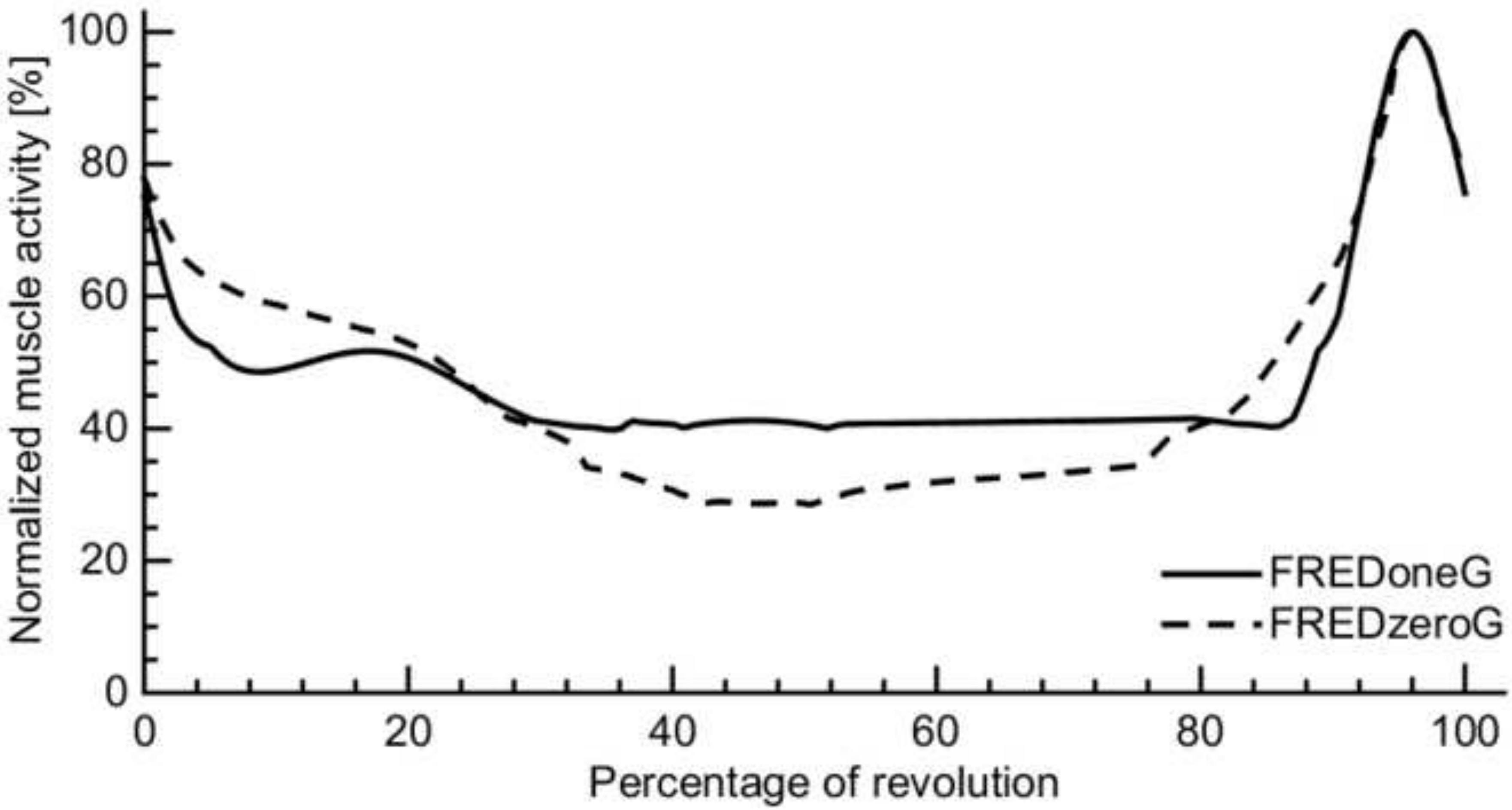

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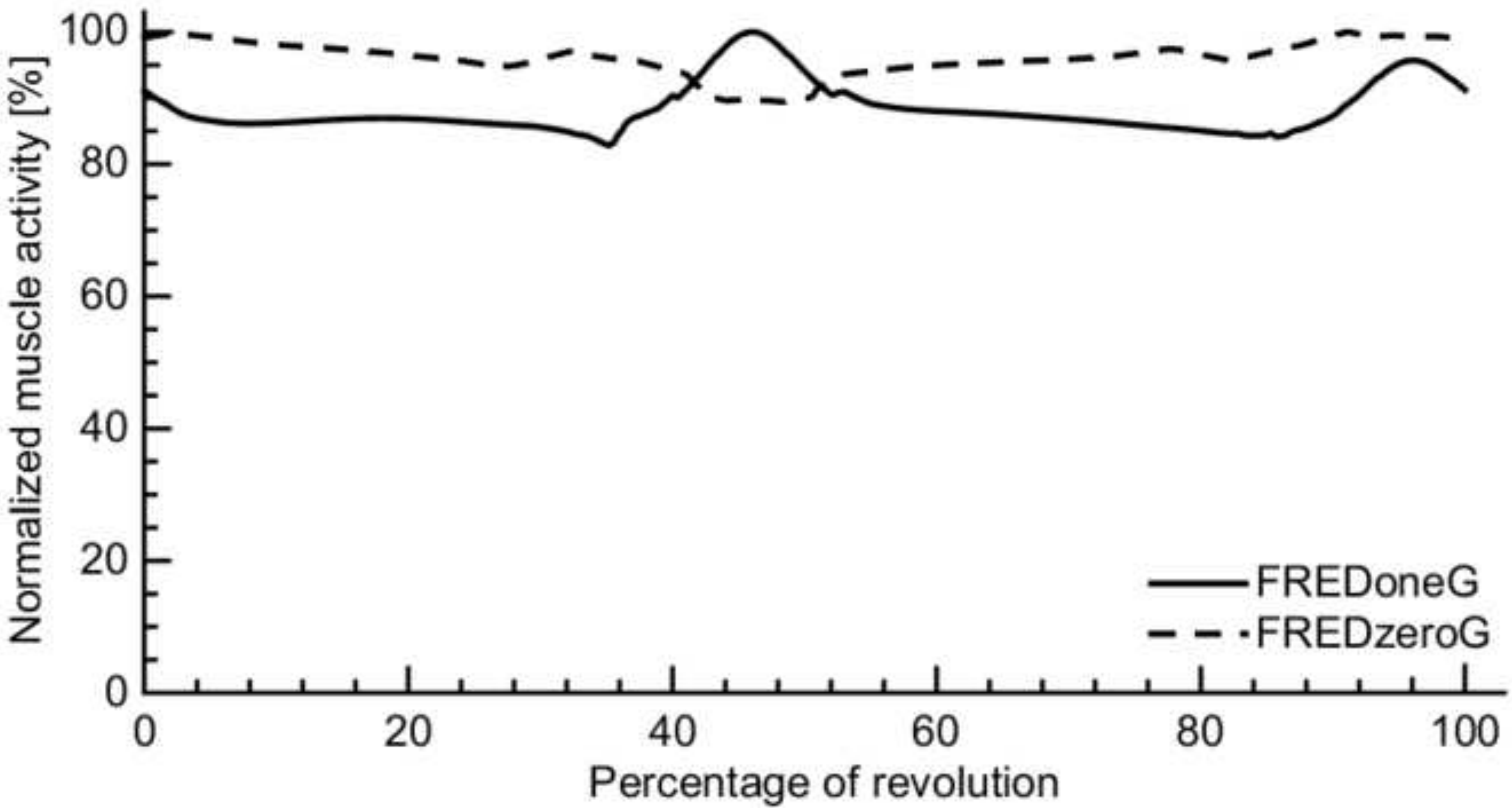

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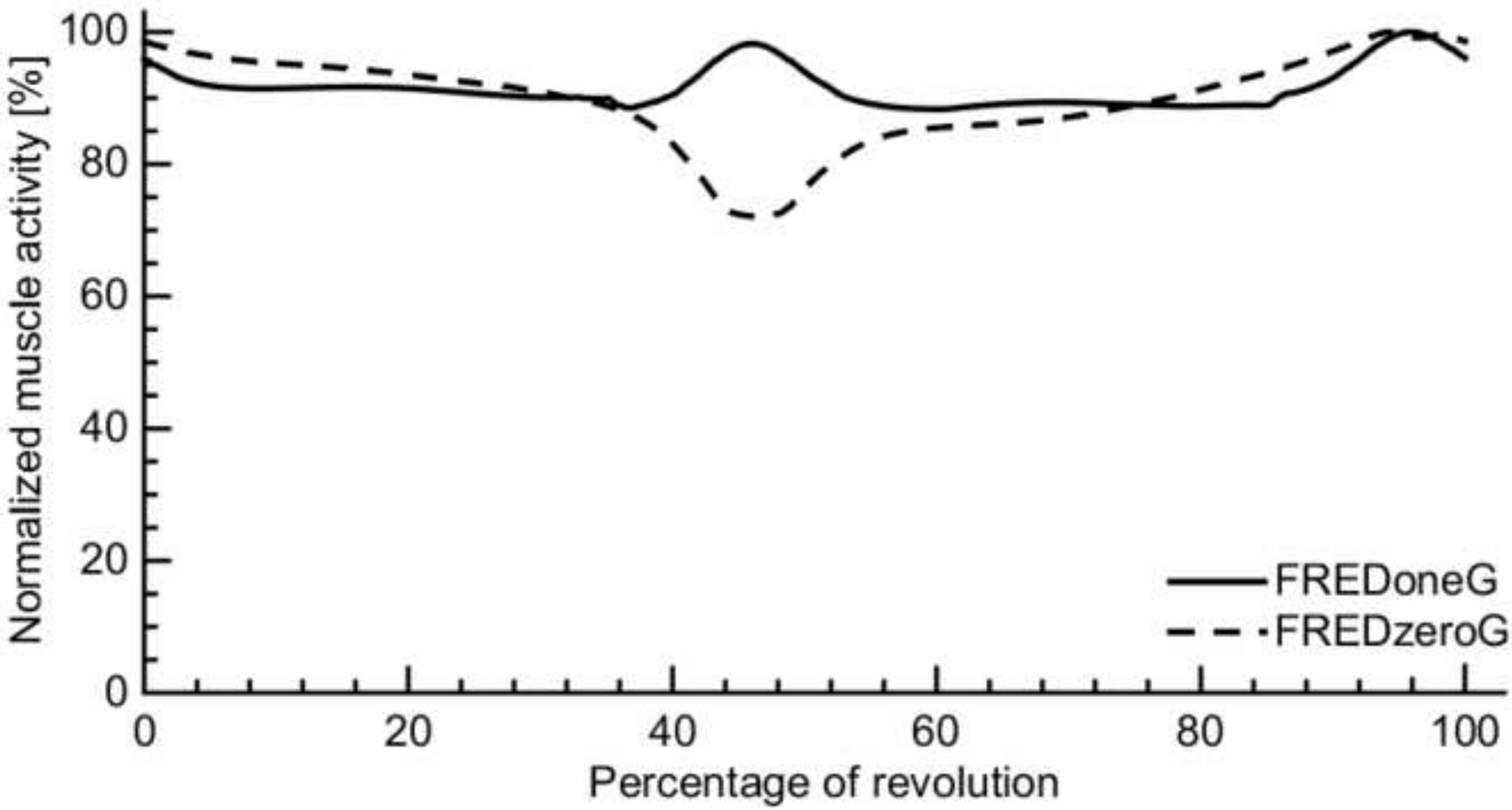

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